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DP-309294

METHOD FOR MONITORING RESPIRATION AND HEART RATE USING A FLUID-FILLED BLADDER

Technical Field

The present invention is related to respiration and heart rate monitoring, and more particularly to a method for monitoring respiration and heart rate based on pressure variation in a fluid-filled bladder disposed in a seat or mattress.

Background of the Invention

Respiration rate, heart rate and their variability are frequently measured as a means of diagnosing and/or analyzing a patient's medical state of health. Such measurements are also indicative of stress level, and a patient is sometimes "wired" to continuously monitor respiration and heart rate during routine or specified situations. It has also been proposed to monitor the respiration and heart rate and the variability of heart rate of the driver of a motor vehicle for purposes of determining the driver's awareness level. Blood pressure and its variability and respiration volume and its variability are also important for analyzing a patient's state of health. Changes in any of these physiological parameters with time may be indicative of a driver's level of awareness, stress, workload or fatigue.

In the case of a vehicle seat, coarse parameters such as occupant weight and presence can be monitored by placing a fluid-filled bladder in or beneath the seat cushion, and measuring the fluid pressure in the bladder; see for example, the U.S. Patent Nos. 5,987,370 and 6,246,936 to Murphy et al., and the U.S. Patent Nos. 6,101,436 and 6,490,936 to Fortune et al., all of which are assigned to Delphi Technologies, Inc. The average fluid pressure in the bladder is proportional to the occupant weight, and variation in the measured pressure as the vehicle is driven can be used to indicate that the occupant is a normally seated child or adult, as opposed to a tightly cinched child seat or infant seat.

Although the bladder-based occupant weight/characterization sensing apparatus is advantageous in that it offers passive and non-intrusive sensing, the information deduced from the pressure measurement has been relatively limited. Accordingly, what is needed is a sensing technique that is passive and non-intrusive in the sense of the seat bladder apparatus, but that is capable of monitoring occupant respiration and heart rate.

Summary of the Invention

The present invention is directed to an improved method for monitoring quasi-periodic physiological functions such as respiration and heart rate using a fluid-filled bladder disposed in a seat or mattress, wherein the bladder pressure is measured and processed to identify minute pressure variations corresponding to the respiration and heart rate of a person that is directly or indirectly exerting a load on the bladder. The respiration rate is identified by band-pass filtering the measured pressure to isolate or extract a pressure component which may be in the range of 0.15-0.5Hz, and the heart rate is identified by band-pass filtering the measured pressure to isolate or extract a pressure component which may be in the range of 2-7 Hz. The extracted pressure components are preferably converted to a digital format, processed and tabulated for comparison with specified thresholds to identify abnormalities and/or anomalies. While the above physiological functions can be characterized by a rate, frequency or periodicity, the characteristics also vary with time, and their variability can be separately measured. This is also true of the amplitudes of the respective pressure components that are related to differential blood pressure and respiration volume. For this reason, the physiological functions are considered to be quasi-periodic.

Brief Description of the Drawings

Figure 1 is a diagram of a motor vehicle seat including a fluid-filled seat bladder and processing circuitry in accordance with this invention.

Figure 2 graphically depicts the AC content of a measured pressure of the fluid in the seat bladder of Figure 1, and two isolated components of such pressure.

Figure 3 is a graph depicting a processed version of one of the components signals depicted in Figure 2.

Figure 4 depicts a representative sampling of heartbeat frequency according to this invention.

Description of the Preferred Embodiment

Referring to Figure 1, the present invention is illustrated in the context of a motor vehicle seat cushion 10 equipped with a fluid-filled seat bladder 12. However, it will be recognized that the invention is not limited to motor vehicle applications, and is applicable to other environments and contexts, such as in a wheelchair, bed, crib, etc. Also, the bladder 12 may be installed under the seat cushion 10 instead of in it, as disclosed for example, in the aforementioned U.S. Patent No. 6,490,936 to Fortune et al., incorporated by reference herein. The components within the region designated by the reference numeral 14 represent the various elements typically present in a vehicular occupant weight sensing system of the type disclosed in the aforementioned patents. In addition to the bladder 12, such elements include a pressure sensor 16 for producing a pressure signal (V_{PS}) on line 18, and a low-pass filter (LPF) 20 for producing an occupant weight signal (WT) on line 22. The pressure sensor 16 detects the pressure of the bladder fluid at a point at or near its center-of-mass. The low-pass filter 20 is designed to remove perturbations of the pressure signal V_{PS} associated with occupant movement and so forth so that the weight signal WT is essentially the DC component of the pressure signal V_{PS}.

Fundamentally, the present invention recognizes that certain perturbations of the pressure signal V_{PS} are associated with quasi-periodic physiological functions of the occupant such as breathing and heart rate, and that such perturbations can be isolated to provide respiration and heart rate information about the occupant. Depending on the mechanical construction of the seat (or mattress, for example), the fundamental heart rate frequency as well

as its harmonics will be transmitted to the bladder 12, the fundamental frequency being in the range of about 0.6 Hz to about 3 Hz. Frequency components above about 10Hz can usually be ignored. Infants and children tend to have heart and respiration rates that are higher than those of adults, and this may require an increase in the monitored frequency ranges. For some purposes, it is desired to determine the pulse-to-pulse interval rather than the heart rate or heart beat frequency.

If desired, the system of Figure 1 may be modified to optimize one or more signal components. For example, the system may include multiple bladders for optimizing physiological information from different locations or to process the various output signals differentially in order to reduce the effects from body movement, vehicle vibration or noise. A single bladder with two or more pressure sensors can also be used for similar purposes since the pressure in a bladder may have spatially local transients. Also, the effects of vehicle vibration or other environmental disturbances can be attenuated and/or compensated for by sensing the presence of such vibration or disturbances with an accelerometer 46, for example. Additionally, the heart and respiration rate components may be optimized by adjusting the base inflation pressure of the bladder 12; to this end, the embodiment of Figure 1 illustrates a fluid pumping system (FPS) 50 coupled to the bladder 12 by a flexible conduit 52. Depending upon the system implementation, measurement of the heart rate and respiration rate components may be optimized with a higher inflation pressure. However, higher inflation pressures may cause the bladder 12 to be too firm for patient comfort. Thus, the optimum inflation pressure will typically involve a trade-off between signal level and patient comfort.

In general, the perturbations associated with respiration and heart rate can be detected by band-pass filtering the pressure signal V_{PS} to identify the signal components in the frequency range of about 0.1Hz – 30Hz or 0.3Hz – 30Hz. The resulting signal V_{AC} is depicted in Figure 2, with a DC offset voltage of approximately 3.5 volts. The relatively low frequency undulation of the waveform is due to the occupant's respiration, whereas the higher frequency undulation is due to the occupant's heart beat.

Referring to Figure 1, the reference numeral 24 designates a band-pass filter BPF_1 for specifically identifying the frequency components of the pressure signal V_{PS} associated with the occupant's heartbeat, and the reference numeral 36 designates a band-pass filter BPF_2 for specifically identifying the frequency components of the pressure signal V_{PS} associated with the occupant's respiration. In the illustrated embodiment, the band-pass filter BPF_1 is configured to pass components of the pressure signal V_{PS} in the frequency range of 2Hz to 7Hz, producing an output signal such as the trace V_{HR} in Figure 2; the band-pass filter BPF_2 is configured to pass components of the pressure signal V_{PS} in the frequency range of 0.15Hz to 0.5Hz, producing an output signal such as the trace V_{RESP} in Figure 2. As with the trace V_{AC}, the traces V_{HR} and V_{RESP} are illustrated with DC offsets so that the traces can be viewed separately. The output of band-pass filter 24 on line 26 is amplified by the amplifier 28 and supplied to an A/D input port of the microprocessor 30. Similarly, the output of band-pass filter 36 on line 38 is amplified by the amplifier 40 and supplied to an A/D input port of the microprocessor 30. The microprocessor 30, which could alternatively be implemented with a digital signal processor, functions to process the input signals to form output signals on lines 32, 34, 42 and 44 representative of the occupant's heart rate (HR), heart rate variability (HRV), respiration rate (RR) and respiration rate variability (RRV). Of course, the microprocessor 30 could also be programmed to compare the depicted outputs with threshold values indicative of normal or marginally abnormal values, and to activate an alarm or warning device when abnormalities or anomalies are detected. Also, it may be desirable to detect changes in the values of HR, HRV, RR and RRV that occur over time for a given individual for purposes of detecting the onset of drowsiness or over-stressing. The same is true of the differential blood pressure (that is, the difference between the systolic and diastolic blood pressures) and respiration volume. The amplitude of the pressure variations due to the heart pulses are also approximately linearly related to the differential blood pressure. The amplitude of the pressure variations due to respiration are approximately linearly related to the volume of

breath exchanged. These physiological parameters and their variability with time can also be monitored as an indication of stress, awareness level, etc.

The signal processing performed by microprocessor 30 to extract the HR and HRV outputs can include local normalization and exponentiation. The signal V_{HR} may be normalized locally according to the following scheme:

$$V_{NORM}(t) = \frac{V_{HR}(t) - V_{MIN}\left(t - \frac{T_w}{2} \le t \le t + \frac{T_w}{2}\right)}{V_{MAX}\left(t - \frac{T_w}{2} \le t \le t + \frac{T_w}{2}\right) - V_{MIN}\left(t - \frac{T_w}{2} \le t \le t + \frac{T_w}{2}\right)} \qquad (1)$$

where V_{MIN} is the minimum V_{HR} signal that occurs in the time interval $\left(t - \frac{T_w}{2} \le t \le t + \frac{T_w}{2}\right)$ and V_{MAX} is the maximum V_{HR} signal that occurs in the same time interval. The time window T_w is selected to be slightly lower than the HR repetition interval, and may be adaptively adjusted if desired. By way of example, T_w may be fixed at 0.8 seconds. In an adaptive configuration, T_w may be reset to 80%-90% of the previously determined pulse-to-pulse duration to ensure that any close-by structured peaks are not confused as heart pulses, while ensuring that the previous or next heart pulses are still counted as heart pulses. Normalizing the V_{HR} signal allows the signal peaks to be easily identified since the peaks all assume a value of unity while the remainder of the normalized waveform has values between zero and unity. The normalization can be further enhanced by raising the locally normalized signal to a power N:

$$V_{NORM-EXP}(t) = \left(V_{NORM}(t)\right)^N \qquad (2)$$

where N = 15, for example. The result of such exponentiation is depicted in Figure 3. Referring to Figure 3, it will be seen that only heart rate pulses remain in the V_{NORM_EXP} signal, and that other perturbations are greatly attenuated. As illustrated in Figure 4, the heart rate HR in beats per minute (BPM) can be easily obtained from either the normalized or normalized-exponentiated waveforms, where HR = 60/Tp, with Tp representing the pulse-to-pulse interval. Heart rate variability HRV may be determined by calculating the

variance of Tp, for example. Alternatively, the microprocessor 30 may perform additional signal processing in the frequency domain (FFT, power spectrum, harmonic spacing, etc.) or the time domain (correlation, adaptive digital filtering, amplification, compensation from other inputs, etc.). In a similar manner, the respiration rate RR may be determined by one of the techniques used for heart rate. If the local normalization technique is used, a larger window size is needed to account for the lower respiration rate. Other schemes such as zero crossing detection could also be used. In some cases, the respiration rate variability (RRV) as well as respiration rate (RR) is of interest; this may be detected in a manner similar to the detection of heart rate variability (HRV).

In summary, the present invention provides a passive, non-intrusive and inexpensive method for monitoring physiological functions such as respiration and heart rate. While described in reference to a human occupant of a vehicle seat, it will be understood that the method equally applies to subjects other environments, and even to non-human subjects that exhibit quasi-periodic physiological functions such as respiration and heart rate.

On an implementation level, it will be recognized that the pressure signal V_{PS} may be transmitted to the detection circuitry by a wireless communication system, if desired, and that the amplifier and filter elements depicted in Figure 1 may be reversed, or the microprocessor 30 replaced with a digital signal processor, as mentioned above. Further, additional band-pass filters may be utilized to detect and monitor body movements, and to detect body movements that are characteristic of choking, convulsions, seizures, coughing, childbirth contractions, etc. The pressure signal V_{PS} and/or the processed HR, HRV, RR or RRV signals may be transmitted wirelessly to a remote site after a vehicle collision in order to assess a medical condition, including whether the occupant is alive or present. In such a case, the presence of the occupant may be determined from the occupant weight signal WT. Auxiliary signals may be included to assist in determining if the vehicle has been over-turned or if the occupant's seat belt is still fastened. Also, the invention may be applied to various types of vehicles, such as aircraft, and to

non-automotive uses such as wheelchairs, bed, cribs and so on. As with automotive applications, a wireless communication could be made to alert medical personnel of an accident condition and assess the medical condition of the subject. Additionally, the invention may involve communications to the subject/patient or another person based on the processed signals, such as a communication that the subject/patient is not moving frequency enough for good health. Moreover, the measured heart and respiration rates can be used as indicators of stress or nervous activity level, from which various conclusions can be inferred; for example, high respiration and heart rate in the case of an aircraft passenger may be used as an indication of extreme nervousness or possible criminal intent. In this regard, it should be understood that methods incorporating these and other modifications may fall within the scope of this invention, which is defined by the appended claims.